High Field Imaging: A Technical Perspective

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The last few years have seen a dramatic rise in interest in high field imaging for both clinical and research applications in humans, with over 200 presentations last year at this meeting, and some published estimates [1] that 1 in 4 "high field" (1.5T and 3T) clinical MR systems sold in the United States in 2005 will be 3T. Higher static field strength presents a number of clear advantages to many applications, but a number of challenges must be addressed as well, particularly related to gradient and RF performance, RF coils, and pulse sequences. The following reviews both the advantages and the challenges from a technical perspective, and highlights areas of opportunity for future development.

In the analytical NMR world, the development of higher field systems (up to 21.1 T [2] and higher) has been driven largely by the need for higher and higher chemical shift resolution. In MR imaging, higher field provides higher sensitivity, provided the associated technical challenges can be overcome. Higher chemical shift resolution provides clear benefits in MR spectroscopy, but plays a less significant role in most imaging applications.

Higher field strength leads to additional effects, some beneficial and some not -- greater magnetization, longer T_1 relaxation times, slightly shorter T_2 relaxation times, greater T_1 dispersion among tissues, increased magnetic susceptibility shifts, and greater B_1 inhomogeneity.

Safety Concerns for High Field Systems

Four general areas of concern have been identified for magnetic resonance imaging [3]: static field exposure, peripheral nerve stimulation due to gradient switching, acoustic noise exposure, and heating due to RF power deposition. Despite years of investigation [4], it has not proven possible to rule out or rule in specific biological hazards due to exposure to static magnetic fields. Based in part on the lack of detrimental effects observed at Ohio State University (8T), the University of Minnesota (7T), and Massachusetts General Hospital (7T), the FDA raised the Significant Risk limit for static field exposure to 8T in 2003, eliminating the need for an Investigational Device Exemption for most studies, and making 7T systems eligible for clearance for marketing under section 510(k) should a manufacturer wish to obtain such clearance.

With regard to peripheral nerve stimulation [5], there is no direct field strength dependence on the stimulation threshold or its effects. As discussed below, however, the requirements for gradient performance increase with field strength due to the need to minimize chemical shift and susceptibility artifacts, and so gradients need to be driven harder and faster. The use of gradient inserts for some applications (particularly in the head) [6] provides the needed performance without increased risk of peripheral nerve stimulation. Peripheral nerve stimulation is produced by the time varying electric field, not the magnetic field, and so depends critically on the physical size of the gradient coil as well as how quickly the gradient field is switched.

Acoustic noise exposure guidelines also are independent of field strength, although the Lorentz forces on the gradient coil windings and other current carrying structures do increase with field strength. A number of aspects of system design -- such as the length of the magnet, the means by which the gradient coil is secured within the bore of the magnet, and the construction of the magnet enclosures -- all have significant influence on the acoustics of the system.

Heating limits themselves are not field strength dependent, but the general decrease in transmitter efficiency means that in practice these limits are reached far sooner. It is often supposed that the specific absorption rate (SAR) scales quadratically with field strength, based on the assumption that this power is deposited as ohmic losses in tissue. Since tissue conductivity generally increases with frequency, this argument would lead to a faster dependence than quadratic, but there is a further problem with this argument -- it assumes that the magnetic field distributions are effectively identical for the two field strengths, leading to identical electric field distributions. Such an assumption is clearly incorrect, and indeed the geometries of coils built for higher field strength are generally quite different from those built for lower field strengths [7], in order to maximize their efficiency over a target volume. While there is generally a decrease in coil efficiency with frequency, and therefore an increase in the power needed to produce a given flip angle, the differences in the volume of tissue over which this power is deposited generally lead to a slower than quadratic dependence of SAR on field strength. A careful study of this trend -- including the optimization of coil geometry for each field strength -- remains to be performed.

Magnetic Susceptibility Effects

Most tissues have roughly the same magnetic susceptibility, about that of water (-9 ppm) [8], although there are some brain structures that accumulate iron [9] and so show a markedly different susceptibility, and the BOLD effect relies on the difference in susceptibility between oxygenated and deoxygenated blood [10]. Susceptibility effects arise primarily due to the difference in susceptibility between tissues and air, such as in the sinuses or the lungs. Because the shapes of the tissues are irregular, the local effects on the magnetic field can have very high spatial order. A further complication in body applications is the presence of embedded metallic objects (like surgical clips) -- even if nominally nonmagnetic, the difference in susceptibility is often great enough to cause local field distortion, and depending on the material conductivity, could cause a local increase in RF heating.

If the local field distortions exceed that produced by the applied gradients, the image will appear distorted. Resistive shims have only a limited ability to compensate for local effects; applying a global correction to compensate a local distortion generally reduces overall homogeneity. In spectroscopic imaging applications, these effects can be quite insidious, producing spurious peaks displaced both spatially and in frequency from neighboring voxels due to the cumulative effects of susceptibility shifts during the acquisition.

The only “solutions” to this problem are stronger gradients (reducing the relative effect of the distortions) and faster acquisition through parallel imaging techniques (reducing the overall phase evolution due to the distortion), although both these approaches have drawbacks and limitations. Gradient strength is limited by the physical size of the gradient coil, the available peak current from the gradient amplifiers, and the maximum gradient slew rate, limited in turn by the design of the gradient coil (its inductance and the induced voltages responsible for peripheral nerve stimulation). Innovations in both coils and drivers are needed to show improvements here; for some applications, local gradient coils may present a solution provided significant mechanical engineering challenges are overcome.

B_1 Homogeneity

High field images typically show “center brightening” due to the inhomogeneous B_1 field distribution within the tissue. Human tissue has both ohmic conductivity, reflecting the motion of free charges in an electric field, and dielectric permittivity, reflecting the reorientation of bound charges in response to an electric field. Both these effects combine to produce the characteristic standing wave patterns, and “center brightening” seen in high field imaging. As noted in [11], a uniform RF magnetic field over an extended volume is simply not a solution to the Maxwell’s equations of electromagnetic fields. It is possible, however, to produce a uniform field over a restricted volume, and this possibility drives some of the development of parallel transmitter systems [12]. The actual field distribution at the higher frequencies encountered in high field imaging is determined often more by the sample than by the conductor geometry in the coil.

Parallel transmitter systems have been demonstrated for head applications but not to date for body applications, and are not currently provided by any of the major manufacturers of clinical MR systems. The basic concept is that rather than providing a single large volume transmitter coil designed to produce as homogeneous a field as possible, several coils -- each isolated from the others -- are individually driven to produce a field which is combined by the object. In general, both amplitude and phase must be adjusted to produce a uniform field, introducing the complexity that at present limits the utility of these systems -- either a single amplifier must divide its outputs among the different elements (requiring a splitter/phase shifter network that can handle high power levels, which is difficult to construct and introduces significant losses of its own), or several amplifiers must be provided. In either case, the amplitude and phase settings of each channel must be optimized for each subject.

RF Coils

For each new field strength, new RF coils are required. To maximize their efficiency as transmitters, their sensitivity as receivers, and minimize their coupling to other coils, the resonant frequency of the coil must match the Larmor frequency of the system, and the impedance of the coil must meet particular requirements (either matching the system impedance for optimum power transfer, or forming part of a detuning network for decoupling). Two aspects of a coil design are the inductance, reflecting the ability of the coil to produce magnetic fields, and the capacitance, reflecting the ability of the coil to produce electric fields. A condition of resonance results at a particular frequency when the inductive energy storage matches the capacitive storage. The inductance is generally determined by the spatial configuration of conductors, and the capacitance is added discretely, although the structure itself has some intrinsic capacitance, which becomes increasingly important at higher frequencies.

For a particular coil geometry, the capacitance needed to resonate the coil decreases as the square of the frequency; when no capacitance is needed, the coil is said to be self-resonant, representing a fundamental limit for the coil. Several groups have turned to “microstrip transmission line” (MTL) coils [13,14], which are designed to be self-resonant, embracing this limitation as a feature. These designs are particularly suited to computer-controlled fabrication and provide great flexibility in the construction of the arrays, and are likely to play an increasingly important role in high field imaging.

A further consideration for the physical size of the coil is related to the volume of tissue to which the coil is sensitive. At higher field strengths, it is generally the case that noise from the subject strongly dominates noise from the RF coil and receiver electronics (indeed, it is generally a design goal to ensure that this condition obtains). Reducing the size of the coil reduces the amount of noise the coil picks up, but also reduces its useful field of view. To recover the field of view, more coils must be added to the array. Higher field systems with more receiver channels will become the norm in the future.

As the coil size is further and further reduced, at some point the noise from the sample becomes less than the noise generated by the coil itself, at which point further reduction of the size of the coil no longer improves the signal to noise ratio. The point at which this condition is reached is strongly dependent on frequency; coil arrays for high field systems will generally need more elements, each smaller in size. Although these smaller elements individually have reduced sensitivity to deep lying structures, contributions from all the elements often combine to give sensitivity comparable to or better than an equivalent volume coil [15].

Contrast Agents

The role of paramagnetic contrast agents is crucial in a number of applications, particularly detecting vasculature changes in tumors. While the relativity of these agents generally decreases with field strength, the effect of paramagnetic contrast agents on signal intensity depends on three factors -- the relaxivity, the T_1 relaxation time, and the intrinsic signal. With the final two factors increasing with field strength, the overall detectability of these agents is expected to increase with field strength, leading to significant improvements in detection of low agent concentration and slow enhancement effects.

Conclusions

The trend to higher field strengths for human MR systems is driven primarily by the desire for sensitivity, but will remain somewhat limited by the need to improve the performance of the gradient and RF chains, and redesign RF coils for higher field operation, as well as the higher cost and siting complexity of higher field magnets. While there are clear engineering challenges to be overcome in delivering the breadth of applications common at lower field strengths, none of these barriers are insurmountable, and new applications enabled by the higher field strengths will arise in the future.

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